A HEART BEAT ACCUMULATOR FOR RESEARCH IN EXERCISE PHYSIOLOGY

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THESIS

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by

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June 1971

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Submitted in partial fulfillment of the
requirements for the degree of

MASTER OF SCIENCE IN ELECTRICAL ENGINEERING

from the

NAVAL POSTGRADUATE SCHOOL
June 1971

ABSTRACT

A system is postulated to count the total number of heart beats in one day. The device is intended for use as a possible indicator of the level of physical fitness of an individual. It is portable, self-contained, and provides for comfortable and natural movement of the subject during the course of daily activities. Electronic design is developed including detailed schematic diagrams of all electronic circuitry involved. The complete plans and photographs of a working prototype are presented. The prototype is to be further tested and evaluated in the exercise physiology laboratory at California State College at Long Beach. It is hoped that this device will prove to be of value in studies and research in physical fitness.

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ACKNOWLEDGEMENTS

The author is indebted to Dr. Gerald D. Ewing of the Naval Postgraduate School Faculty for his continuing interest, guidance, and encouragement throughout the development of the prototype unit. The author also wishes to express his appreciation to Dr. Joseph A. Mastropaolo, Director of Research in Men's Physical Education at California State College at Long Beach for defining the possible need for such a device, and for his continued assistance throughout its development.

I. INTRODUCTION

Biomedical engineering is a field that has attained increased notoriety during the past decade. An indication of the interest in this area to the engineering community has been the formation and expanded publication of the Institute of Electrical and Electronic Engineers' Transactions on Biomedical Electronics. Electronic applications to the numerous areas of biology, physiology, and related disciplines seems illimitable, since the human body is such a complex system, so far from being completely understood.

One such associated field is exercise physiology of which physical fitness is an important segment. At present, there are numerous theories and methods concerned with measuring levels of physical fitness. The total number of heart beats over a twenty-four hour period may be another technique to determine the level of physical fitness of an individual. Presently, however, no device exists which is self-sustaining, totally portable, and capable of counting the total number of heart beats in a day.

Such a device must be small, lightweight, as well as self-sustaining. Furthermore, it would have to be placed on the body in such a manner that would not impede the movement of an individual during the course of daily activities. Accuracy is another requirement, although total precision would not be a necessity since an error of ten heart beats

over a twenty-four hour period would be negligible. Realistically, low cost design is also a certain requisite.

This thesis is directed towards designing, testing, and providing a prototype of such a device.

II. SYSTEM DESIGN

A. LINEAR OR DIGITAL

Determining a specific electronic design required that certain basic questions be resolved concerning the overall system. The project indicated the use of either a linear or digital system. After considering the desirability of each, the decision was made for a digital system. This decision was based on the accuracy requirements of the proposal. It is impossible for a linear unit, such as the voltage increase on a capacitor, to be as accurate as an actual counting device, since there are natural limitations to the number of significant figures on all linear readouts. By adopting a digital counting scheme, all accuracy requirements are met inasmuch as it is an exact system.

B. LOCATION OF ELECTRONICS

The second basic decision involved the placement of the electronics on the human body in order to effect the minimum amount of discomfort and impedance to movement. Ideally, a device the size of a wristwatch would be desired. However, after consideration of the electronic parts that would in all probability be involved, it became obvious that a device of this size would not be feasible without integrated circuit design. The electronic components dictated that the size of the device be such that it would most effectively be

placed on the trunk of the body. Consideration of the various possibilities indicated the waist would serve as the most effective area for placement. Around the waist a belt of electronics could be worn, while not inhibiting any movement of the limbs, thus providing for virtually natural movement. This decision resulted in defining the approximate allowable area for the electronics.

C. INPUT DEVICE

The choice of input device was the next consideration. There were two good possibilities at the outset - a pressure pulse transducer, or direct electrical body contacts (electrodes). Both offered various advantages and disadvantages. The pulse transducer had one serious weakness which dictated the use of electrodes as the input device. The pulse transducer, because it is a pressure device, is far too susceptible to noise signals generated by the body. If a pulse transducer moves on the skin, erroneous output pulses may easily result. It would be virtually impossible to attach a pulse transducer to an exercising subject, and still limit noise due to transducer movement without causing extreme discomfort to the individual involved. Thus the electrode became the logical choice, for exercise and movement are certainly part of the daily routine.

Furthermore, during the electronic design of the system, further advantages of the electrode were discovered and will be discussed later.

Once the electrode was chosen as the desired input device, it was necessary to determine what type of electrical signal which would be

picked up on the skin by the electrodes. This signal would dictate what type of signal processing might be necessary.

Electrically, the heart is basically an oscillator. Prior to each contraction of the heart, an electrical impulse is set up by the sino-atrial node and is propogated from there throughout the heart. As the electrical signal passes through the heart (triggering the mechanical contraction), tissues around the heart also receive "leakage signal." This leakage is transmitted to the surface of the skin, but with significant attenuation. By placing electrodes on the skin on opposite sides of the heart, the generated electrical signal can be observed. (For a more detailed discussion, see Ref. 1). A typical record of these potentials, known as the electrocardiogram, is shown below:

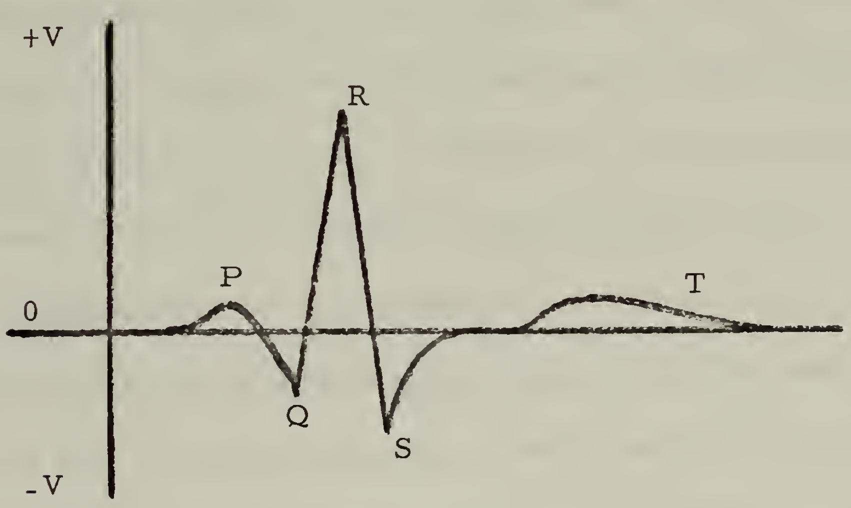

Fig. 1 Electrocardiogram

The "P" wave is caused by electrical currents formed as the atria depolarize; "QRS" complex by currents from the ventricles prior to contraction; "T" wave by currents formed as ventricles recover.[1]

Figure 1 is drawn approximately to scale. As is shown, the "R" sector of the "QRS" complex is by far the most prominent pulse. Because of its prominence, the "R" portion was chosen to be counted in the system. Because of noise, and the non perfect shape of the signal, it became evident that some form of pulse shaping would be necessary.

D. OUTPUT UNIT

The final basic question to be resolved concerned the output device. Was it necessary to require the output unit to be connected at all times to the counter, or should it be added just for readout purposes? Since readouts would be observed for a short time relative to the period of actual pulse counting, it seemed senseless to require that the output device be worn at all times. Furthermore, by connecting the device only when readouts are desired, reduction of the size and weight of the worn apparatus was accomplished. However, requirements for a small and self-sustaining output unit were not eliminated. The subject must be able to easily transport the output unit when desired.

Thus the overall system design was completed. It would be a digital based system using electrodes as the input device, with the

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[1]Guyton, Arthur C., Textbook of Medical Physiology, p. 196.

output unit separated from the functioning system until readout is desired. Because of the electrode output, pulse shaping would be necessary. A block diagram of the system is shown on the following page.

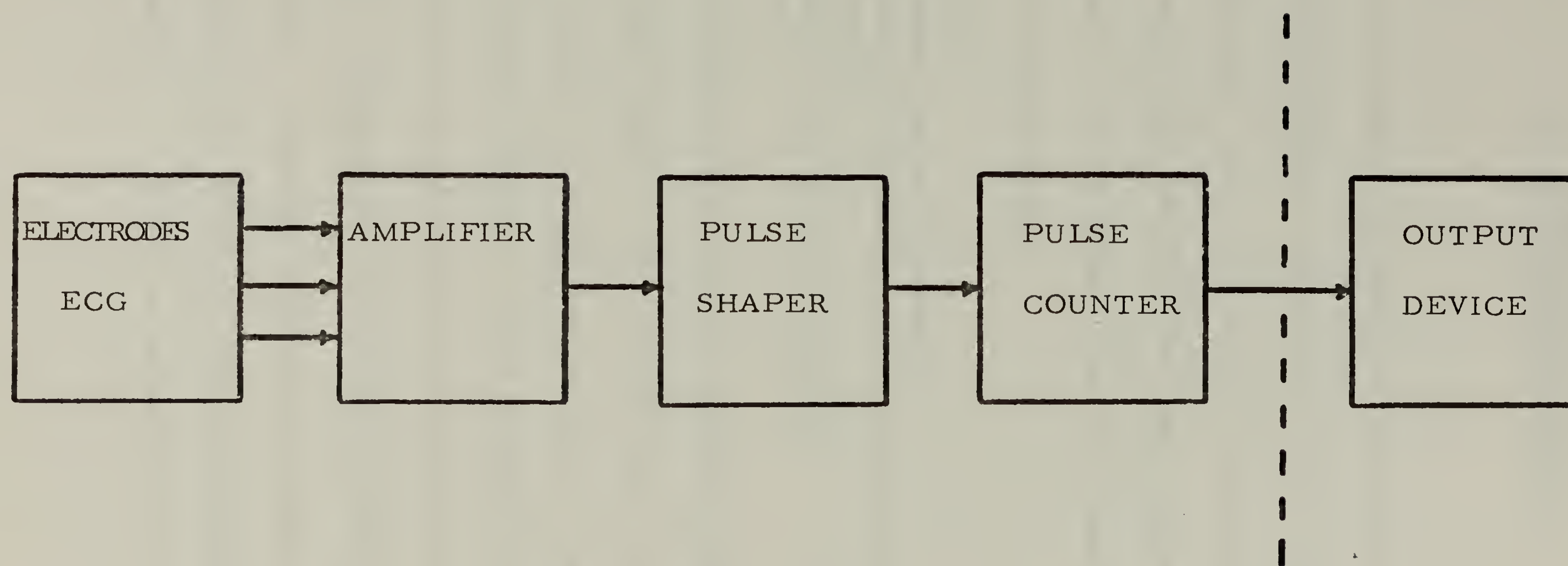


Fig. 2 Block Diagram of Heart Beat Accumulator

III. ELECTRONIC DESIGN

A. COUNTER

The electronic system was designed around the digital counter. Because it was the foundation of the device, the counter unit was the first to be considered. As the basic unit, the various specifications of the counter would determine the electronic requirements for the other components. It was postulated that the counter satisfy the following general requirements:

1. Small size
2. Low-power; low supply voltage
3. Availability of all outputs (if integrated circuits)
4. Able to reset
5. Acceptable loading characteristics
6. Low cost

Integrated circuits were indicated in order to best satisfy the limiting size requirements. After considering the various integrated counters, the Fairchild CuL9989 was selected. A logic diagram of the CuL9989 is given in figure 3. As is indicated, each unit consists of four cascaded flip flops (or four binary bits) with a readily accessible reset line. Furthermore, one standard load for the CuL9989 is defined as itself. Therefore, by merely connecting 0_4 to the count input of another CuL9989, more bits can be obtained without providing for external circuitry. Any desired number of bits could thus be obtained.

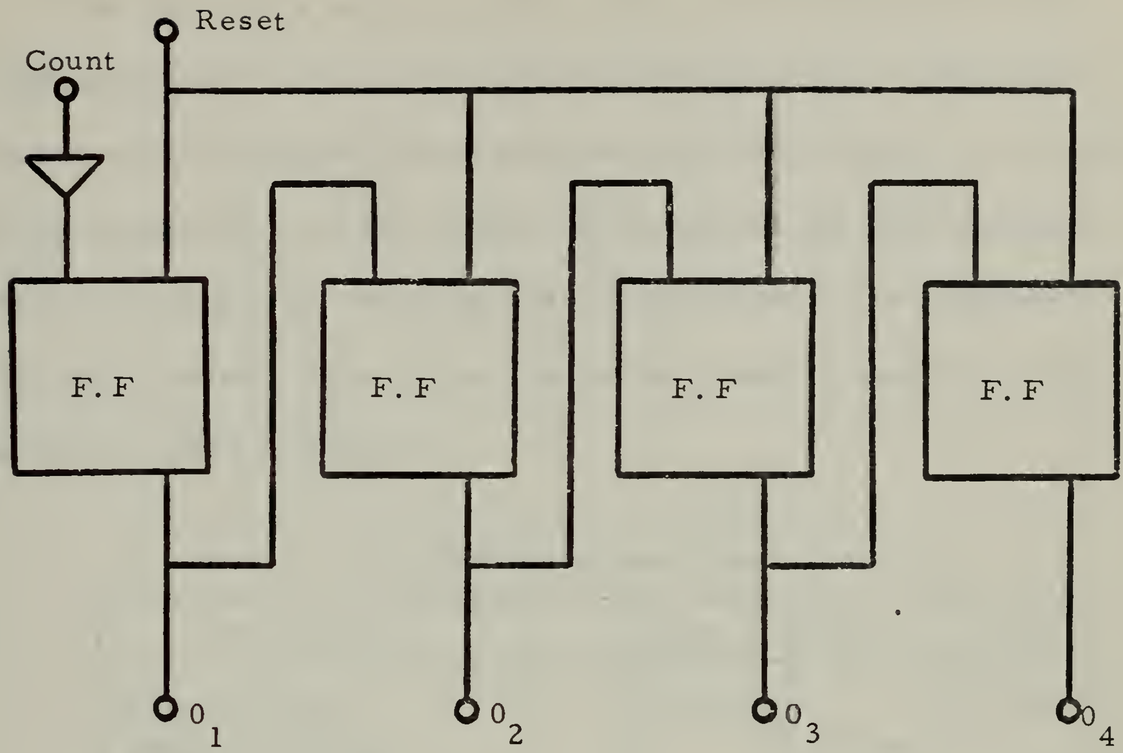


Fig. 3 Logic diagram of the CuL9989

In order to determine the desired number of bits, the count capacity of the unit had to be estimated. Reference 1 indicates that 75 is the number of heart beats in one minute for the average individual. The worst case situation of 100 beats per minute over a 12 hour period would produce 72,000 beats. By supplying 16 binary bits, a counter would be capable of ($2^{16}-1$) or 65,535 counts. Since a digital counter recycles to zero, and continues to count, 16 bits would provide the necessary count capacity. It was assumed that one reading would occur at least once during a 12 hour period. Four cascaded CuL9989's provided the necessary 16 bits.

Four CuL9989's were acquired. They were cascaded by direct coupling between units, and found to function perfectly. These four units provided 16 binary bits without any external circuitry in a volume of approximately two cubic inches. This became the digital counter, and the foundation for the remainder of the device. The pertinent electrical characteristics to be used as the requirements for further design are given in Table 1.

Table 1. CuL9989-Electrical characteristics

Parameters	Min.	Typ.	Max.
Supply voltage	3.6v		5.5v
Power dissipation		132mw	
Count input-low			0.45v
Count input-high	1.2v		
Count input current		210uA	
Count input pulse width	40ns		
Count input slope-pos. going		1.0v/us	

B. ELECTRODE

The Beckman biopotential skin electrode was selected as the most suitable electrode investigated. It is easily applied, not affected by noisture, and provides signals free from motion artifacts. Furthermore, it can remain in position for days. Beckman has had the electrodes on subjects for as long as six days while receiving clean signals. This electrode appeared ideally suited for the system.

The input system would be supplied by three electrodes as is standard for the electrocardiogram. With two electrodes placed

across the heart, and another lower near the waist as the common connection, an equivalent electrical circuit was postulated (figure 4).

When properly applied, the electrode has an impedance of approximately 1K (Ro). All other resistances are tissue originated, and thus quite small. The output signal is in the low micro volt region.

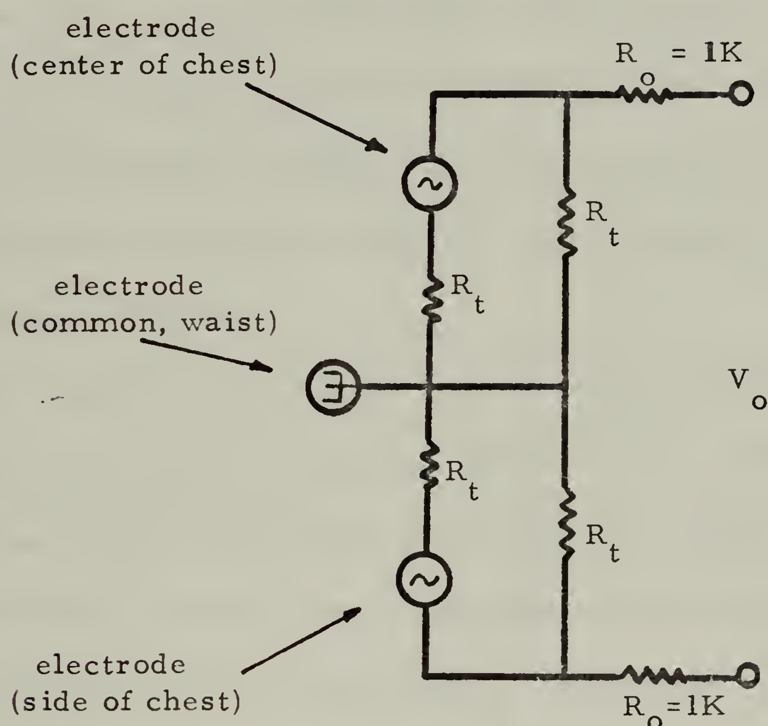


Fig. 4 Electrode Equivalent Circuit

C. AMPLIFICATION

By observing the electrode equivalent circuit, it was apparent that a differential amplifier was indicated. Not only was the differential input suitable for the electrodes' signals, but all unwanted noise common to the electrodes, such as the common 60 cycle, would be eliminated without the need for filtering. A good differential amplifier

with a high common mode rejection would produce the desired amplified signal. A voltage gain of approximately 100 dB. would be required to attain a one volt signal. The operational amplifier, with the inherent high gain, differential input, and respectable common mode rejection was the logical choice.

The problem was to find an operational amplifier with the required high gain while operating with supply voltages between 3.6 to 5.5v as dictated by the digital counter requirements. The writer was unable to locate an operational amplifier with 100 dB voltage gain which would operate at such low supply voltages. It was therefore necessary to cascade two amplifiers.

The RCA 3029A operational amplifier was selected as the best available unit capable of operating with these low supply voltages. The 3029A has an open loop voltage gain of 60 dB. However, due to stability considerations, the open loop configuration was not desired. Feedback near open loop gain was thus utilized. The schematic of the amplification stage is given on the next page (figure 5).

It was necessary to capacitively couple stages to insure that no dc level would reach the second stage. With the high gains involved, any small variation in the dc offset from the electrodes might cause a significant dc level to appear at the output. Because the electrode placement is slightly variable, it was impossible to permanently balance the first stage, hence the need for the capacitor. Although the heart has a pulse repetition rate of approximately one pulse per

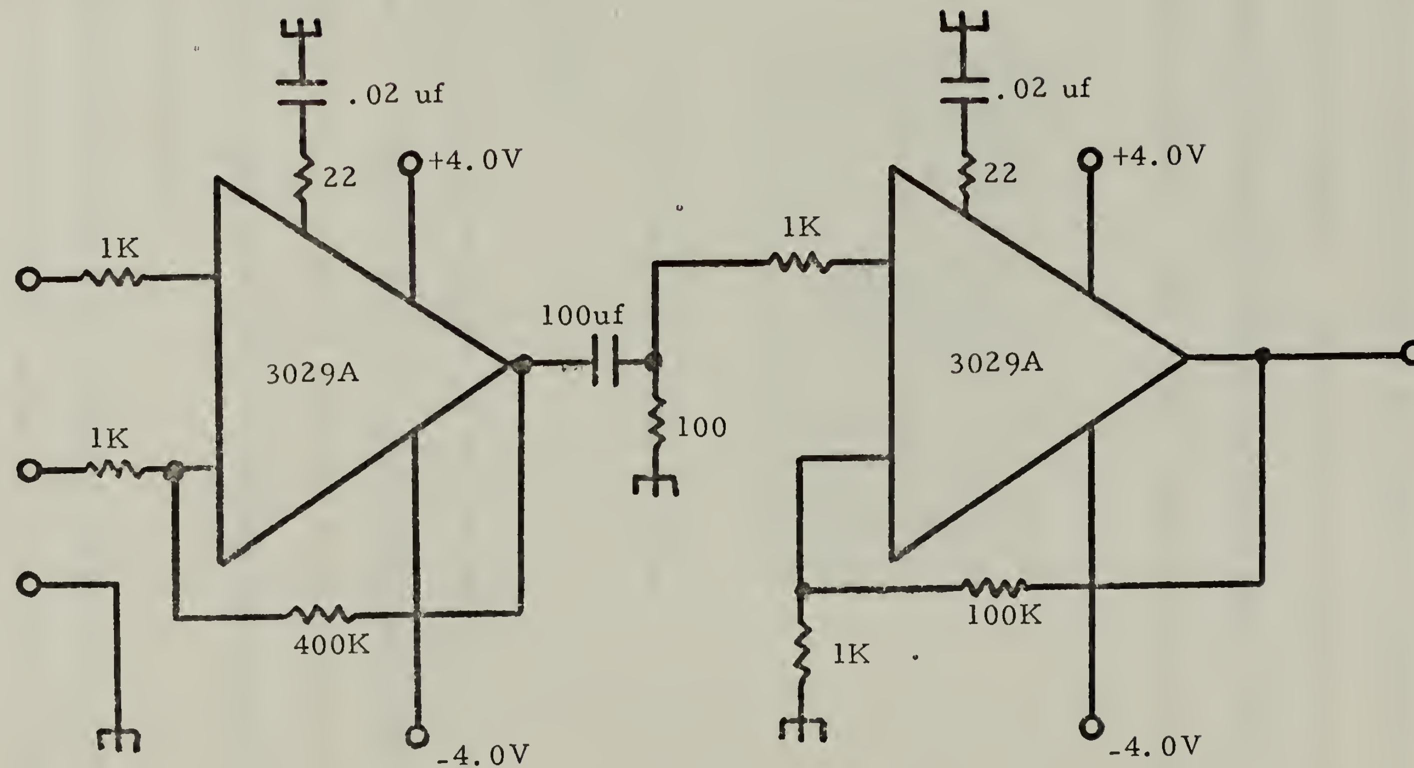


Fig. 5 Amplification Stage

second, the frequency components of the pulse are much higher and easily passed by the 100 uF capacitor.

The output of a heart beat after it has passed through the amplification stage is shown in figure 6. It is very interesting to note the striking similarity between the actually recorded pulse with the text-book pulse shown in figure 1. The "QRS" complex is very apparent.

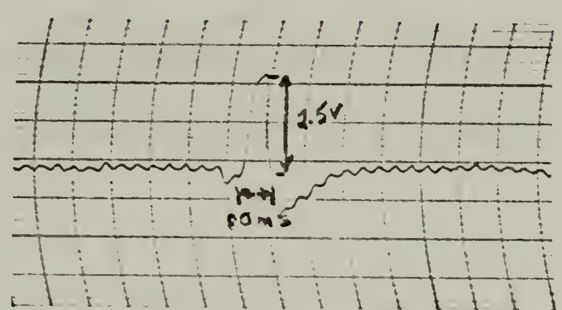


Fig. 6. Amplified Heart Pulse

D. PULSE SHAPING

The amplified heart pulse (figure 6) was not suitable to trigger the digital counter. The pulse has more than sufficient duration, but the requirement for a positive going signal of 1v/us was not supplied. At best, the pulse has positive slope of 1v/10ms. Some degree of pulse shaping was indicated.

A monostable multivibrator or a Schmitt trigger were the two logical choices. No integrated one-shots were available, while a Schmitt trigger could be fashioned through simple modification of an integrated differential amplifier. Therefore, due to the size of an

integrated circuit relative to a discrete component unit. the Schmitt trigger configuration was preferred. The RCA CA3000 was selected primarily due to its low supply voltages and immediate availability. Figure 8A exhibits the Ca3000 modified as a Schmitt trigger as given in Ref. 5. The trigger levels are controlled by the 250K potentiometer. These levels were adjusted for approximately one volt, well removed from the noise levels present in this position of the circuit. This corresponded to a 100K potentiometer setting. The Schmitt trigger provided a nearly perfect square wave as depicted in figure 7.

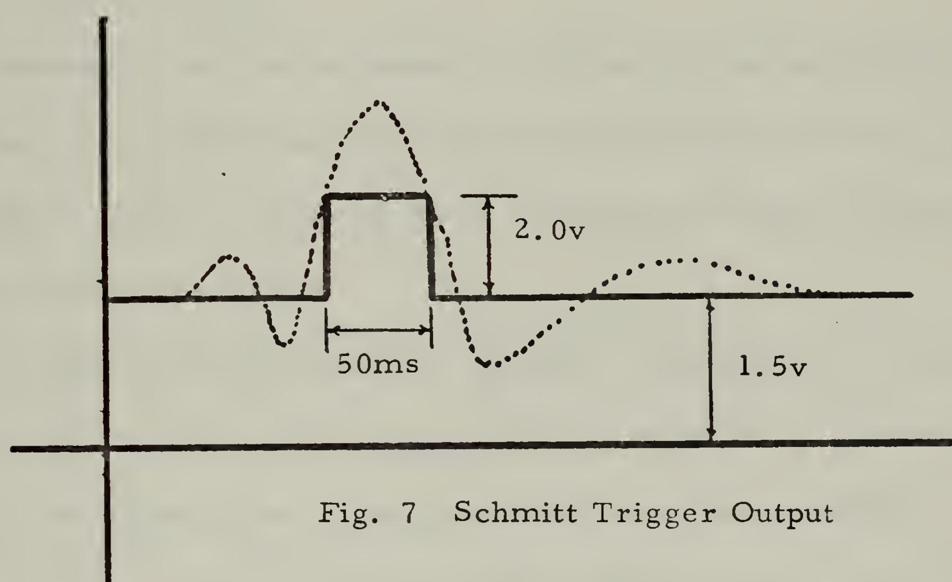


Fig. 7 Schmitt Trigger Output

As is apparent from figure 7, there was a definite dc level present in this output. This dc level would cause the counter to remain in a constant state. By utilizing silicon diodes, this level could easily be removed without any significant increase in size of the stage or altering of the pulse shape. The positive going slope of the pulse more than met the required lv/us of the counter.

The single-ended output impedance of the CA3000 is approximately 10K, while the input impedance of the counter is 3K. In order to match these two impedances to prevent distortion of the Schmitt trigger output, an emitter follower circuit was indicated. Since only a positive pulse was desired (and present), no biasing was required on the base of the emitter follower. Furthermore, one diode drop (removal of dc level), V_{BE}, from the silicon npn transistor was added. Figure 8B shows the addition of the emitter follower circuit.

E. OUTPUT DEVICE

Because it was not necessary for the subject to understand the output data, the simplest and most inexpensive method of readout was indicated. Hence a binary array of incandescent lamps was selected as the most effective method.

In order to avoid loading the counters excessively, a transistor switch was dictated. The lamps selected, Muralite, are rated at 10v, 20mA, the lowest power consumption for incandescent lamps that was encountered. As the switch, the silicon npn transistor 2N3417 was selected because of its immediate availability and relatively high gain. Since the counter output (high 1.2v, low .45v) would "switch" the transistor, no biasing was required. A 10K base resistor was used to limit base current. The basic transistor switch is given in Figure 9A. The base current is approximately 60uA and does not excessively load the counter.

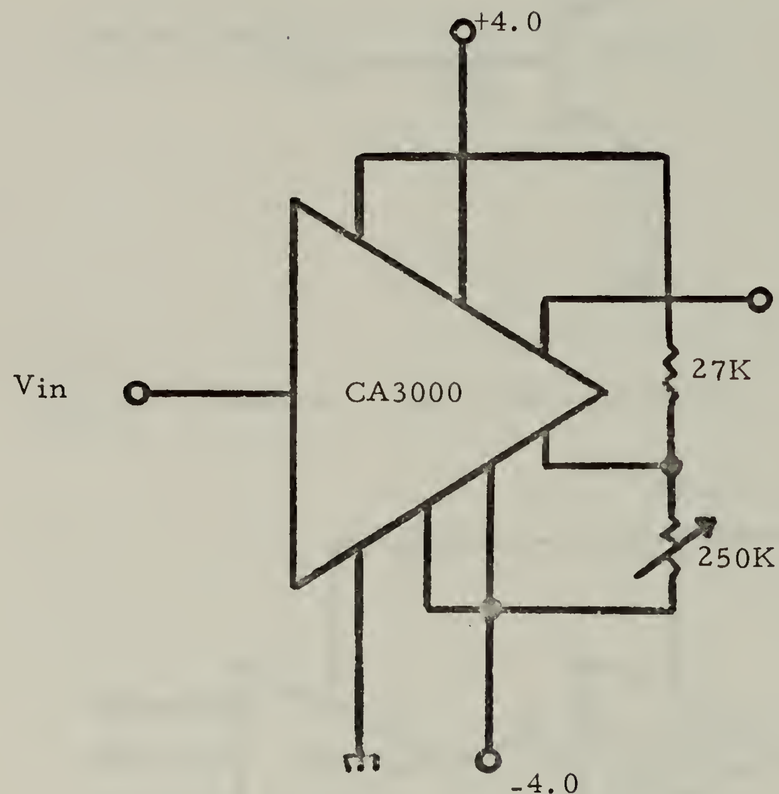


Fig. 8A - Schmitt Trigger

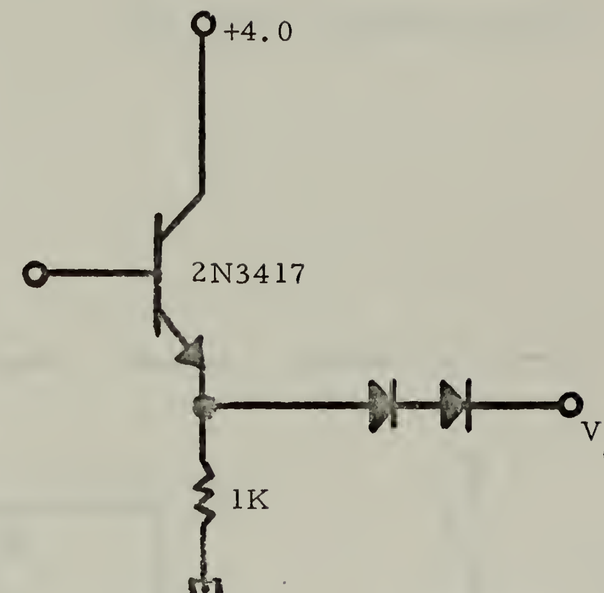


Fig. 8B - Emitter Follower

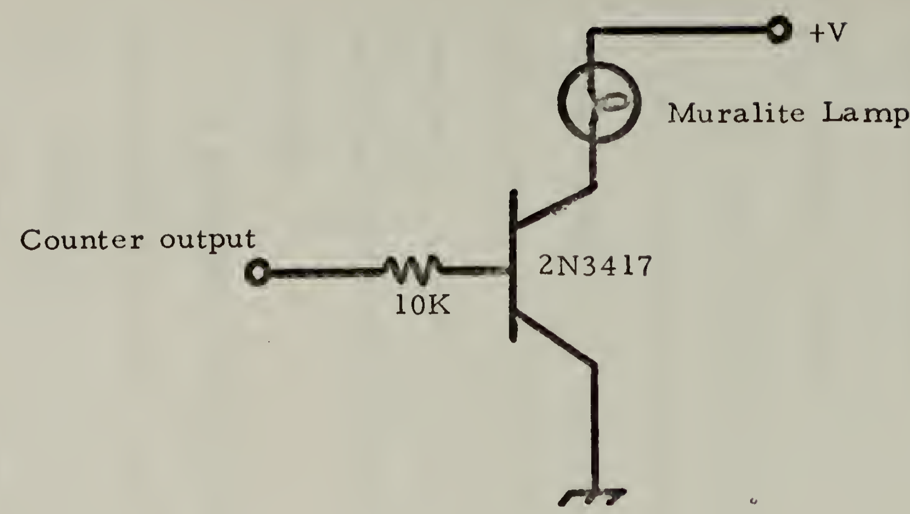


Fig. 9A Transistor Switch

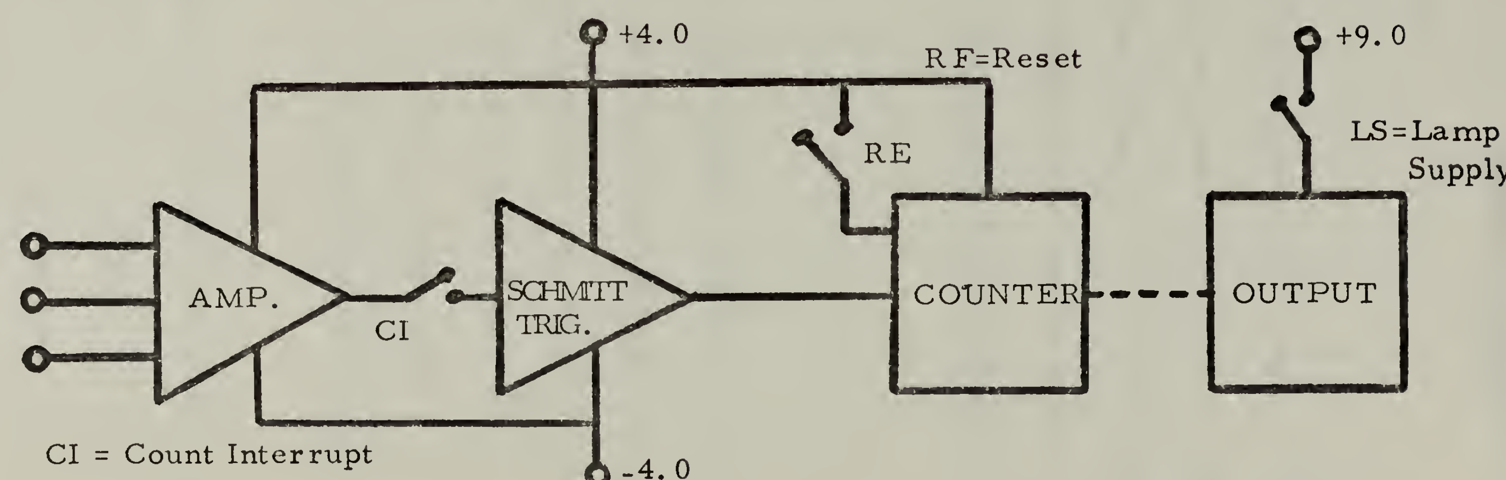


Fig. 9B Switch Locations

The output configuration dictated that three manual switches be inserted in the system. These switches were required for the following:

1. Turn power on to indicator lamps.
2. Interrupt count for readout.
3. Reset digital counter.

Figure 9B indicates how the switches were placed in the system. The switch to interrupt the count was placed just prior to the Schmitt trigger since triggering of the counter had the least chance of occurring from any noise generated by the switch. The input to the counter is very sensitive, and switch noise could easily trigger the count if placed directly on the input.

F. POWER SUPPLY

The entire system was designed to have supply voltages between 3.6 to 5.5v with the operational amplifiers requiring both positive and negative supplies. With the device being portable, the smallest batteries capable of supplying the necessary power would be the most desirable. The power dissipation by the counter (approximately 1/2 watt) was by far the largest and it thus set the requirements for the batteries.

It was hoped that hearing aid batteries, with their extremely small size, might be capable of delivering such power. After investigation, this was found not to be the case. With size still the primary factor, alkaline and mercury batteries were tested since both types are designed for long life and continuous drain. After careful consideration,

the mercury batteries were determined to be the best suited for the desired power unit. Mercury, while loaded, essentially maintains a constant voltage and then drops to zero when its useful life is terminated. Alkaline voltages decay with use, and do not supply a constant voltage. Since both positive and negative supplies should always be of approximately the same magnitude, and in this device these supplies would not be equally loaded, mercury was indicated.

Four Eveready E133N (4.05v), three in parallel for the positive supply, and one for the negative supply, became the power unit. Three batteries for the positive supply were necessary to meet the power requirements of the counter. The common 9v transistor battery was determined to be most suitable to power the indicator lamps of the output device.

IV. THE PROTOTYPE MODEL

A. CONSTRUCTION

After the completed design was tested in the bread board configuration, it was necessary to construct a prototype model for testing under working conditions. The circuits were wired on vector boards, with each stage mounted on a separate piece of board approximately 1" x 1.5", with wire connections between each stage. The reason for utilizing separate pieces was to insure maximum flexibility since the electronics would be worn around the waist. After wiring the prototype and preliminary testing in the laboratory, all connections were lacquered to protect against shorts.

The output device was constructed in a box, 3"x5"x1.5. Included in this unit were the switches for readout, clear, and reset. Furthermore, all transistor switches were added to the output unit in order to minimize the size and weight of the actual device which would be worn by the subject. Photographs of the prototype are given in Appendix A.

Because all packaging and fabricating were performed by the writer, the prototype is in crude form. However, it is functional and does provide a testable model.

B. PRELIMINARY TESTS AND MODIFICATIONS

In order to determine if the prototype model was functioning as designed, it was necessary to perform certain preliminary tests. The

electronics and batteries were taped to a belt which was worn by the writer for a period of five minutes. By averaging the readout for this period and comparing this with the pulse rate of the subject it could be determined if the device was working properly. After numerous tests, the device was found not to be functioning correctly.

The problem was loading the device while the counter was in a dynamic state. Even if the individual lamps were not connected to their supply voltage, the connection of the output device while the counter was counting caused erroneous results to occur. This was determined to be the problem, since if the output device remained connected at all times, the count was correct. Furthermore, if the count was stopped, and the output device disconnected and then reconnected, the readout remained the same. The solution appeared to be to interrupt the count before the output device would be connected to the system. The unit was modified by placing the count interrupt switch on the belt rather than on the output device as originally intended.

The device was tested as before on a resting subject by obtaining the average pulse rate from the device and comparing the result with the pulse rate of the subject. Since under conditions of fatigue the heart signal might be somewhat altered, it was necessary to test the unit under these conditions as well. The device was placed on an exercising subject (heavy breathing and deep inspiration) and tested with the comparison of pulse rates used previously. Under both conditions of rest and exercise, the unit performed correctly.

V. CONCLUSIONS

The preliminary tests indicated that the prototype model does function properly, and provides a useful device for further testing and evaluation. Furthermore, the electronic design of the system was verified through this testing. The cost for materials for the prototype was approximately $100.00, with the digital counter and electrodes accounting for 75 per cent of this figure.

The major shortcoming of the prototype is the continuous power required. If the primary consumer of power, the counter, could be replaced by a lower power unit, the size and number of batteries could be reduced resulting in a significant improvement in size. The RCA CD4004D seven-bit low power binary counter might be such a replacement. The CD4004D has a typical quiescent power dissipation of 5uW while the power dissipation of eight binary bits utilizing CuL9989's is approximately 250mW. With the replacement of the CuL9989 counter by a lower power unit such as the RCA CD4004D, and utilizing industrial technology, the overall size of the device could easily be reduced to at least 25 per cent of the prototype size.

The device is to be donated to the exercise physiology laboratory at California State College at Long Beach for further testing and evaluation. It is hoped that the device will prove to be of value in studies and research in physical fitness.

APPENDIX A

PHOTOGRAPHS OF PROTOTYPE

Figure A-1 shows prototype in position on a subject

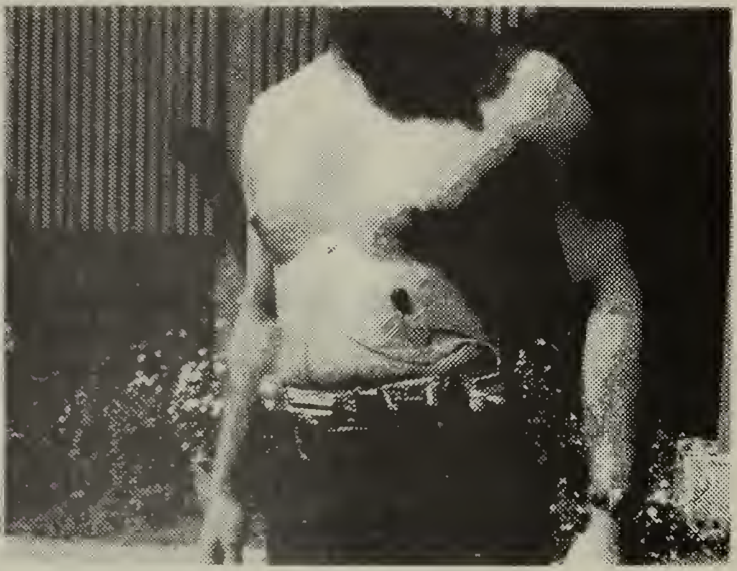


Figure A-1

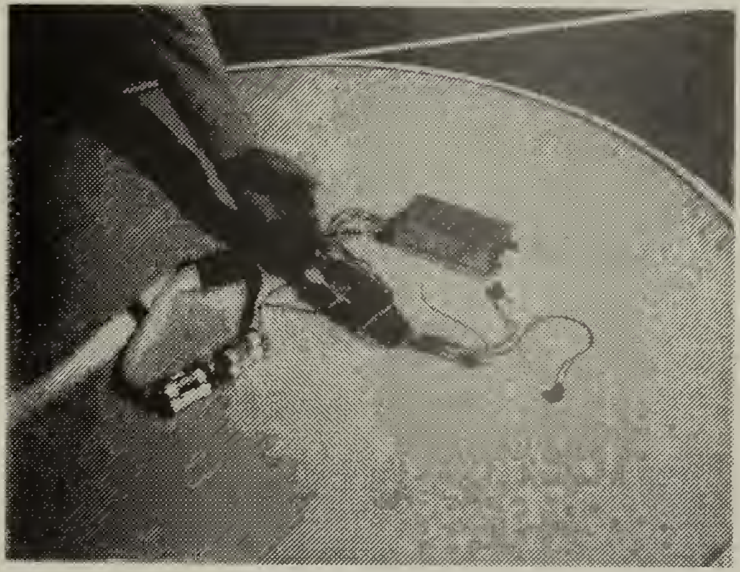


Figure A-2 diaplays electronics, output unit, and electrodes.

Figure A-2

APPENDIX B

OPERATING INSTRUCTIONS

APPLICATION:

1. Fit electronic belt around waist.
2. Attach electrodes to body (black, near waist; white, near center of chest; red, on side of chest) utilizing Beckman adhesive collars and electrode paste as indicated below:
   a. Clean skin with 70% alcohol.
   b. Remove paper from one side of adhesive collar and press onto the electrode.
   c. Apply electrode paste into holes on electrode face.
   d. Remove paper from the other side of the adhesive collar and apply to the skin.

READOUT PROCEDURE:

1. Stop count by switching the count interrupt switch located on the electronic belt.
2. Connect readout device.
3. Switch lamp supply switch on (LS).
4. Record all lamp numbers which are <u>not</u> on.
5. Reset counter by pressing "RE" switch.
6. Turn off lamp supply switch (LS).
7. Disconnect readout device.

8. Resume count when desired with switch located on belt.

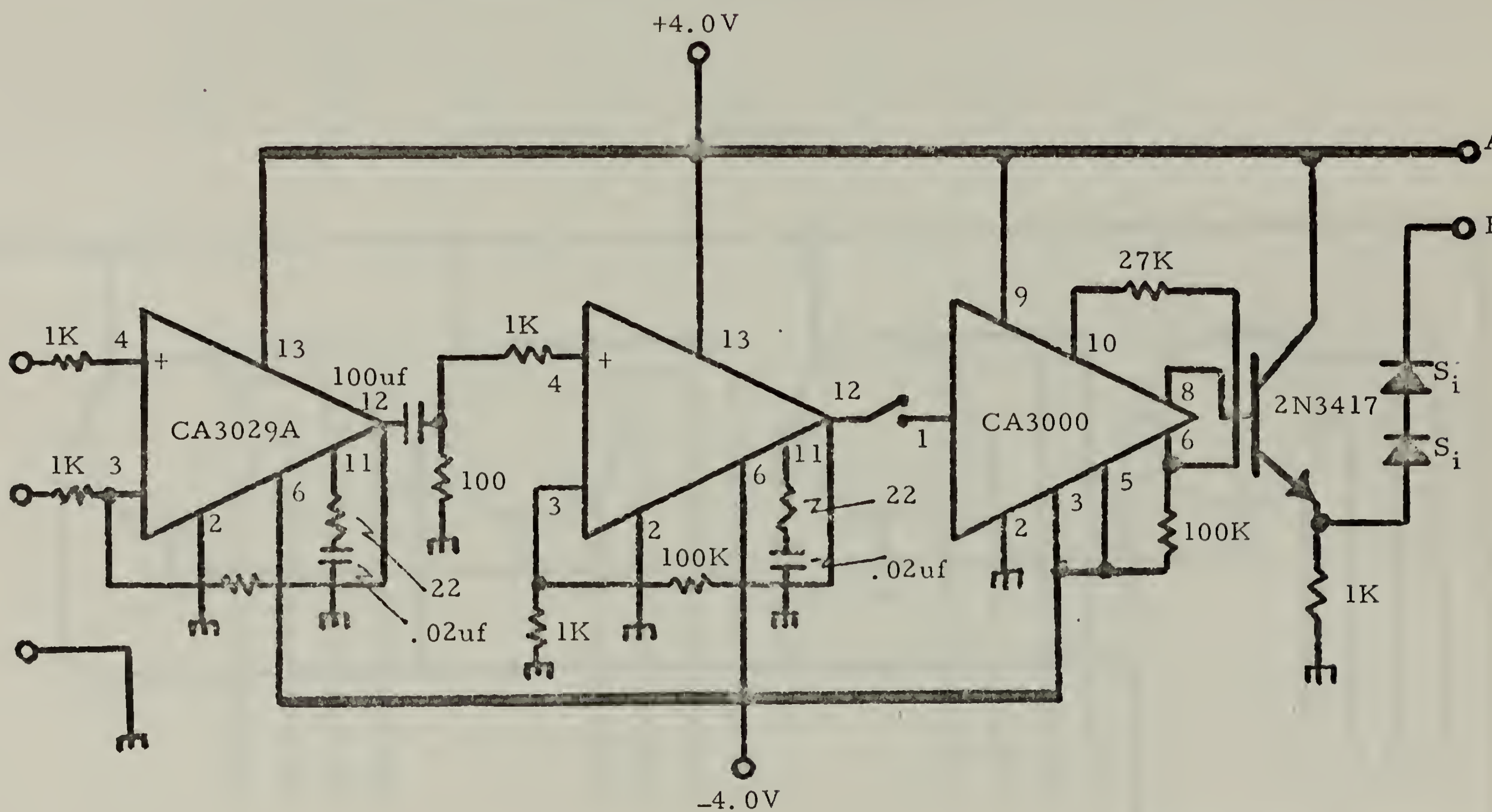
9. Refer to Table B-1 for interpretation of data.

TABLE B-1

Lamp Number	Decimal Equivalent
1	1
2	2
3	4
4	8
5	16
6	32
7	64
8	128
9	256
10	512
11	1024
12	2048
13	4096
14	8192
15	16384
16	32768

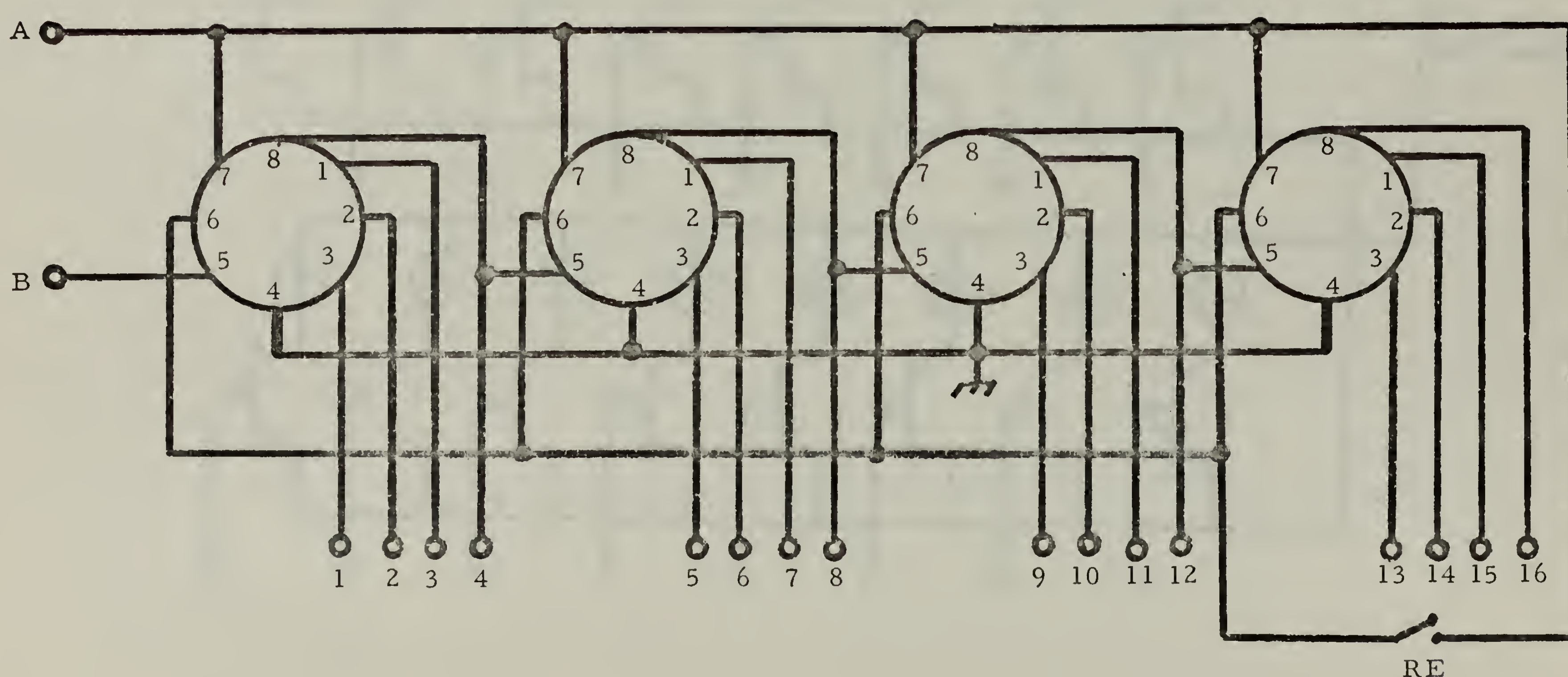
Take each recorded lamp number and write down its respective decimal equivalent. Add all decimal equivalents. This is the total number of heart beats.

AMPLIFICATION AND PULSE SHAPING



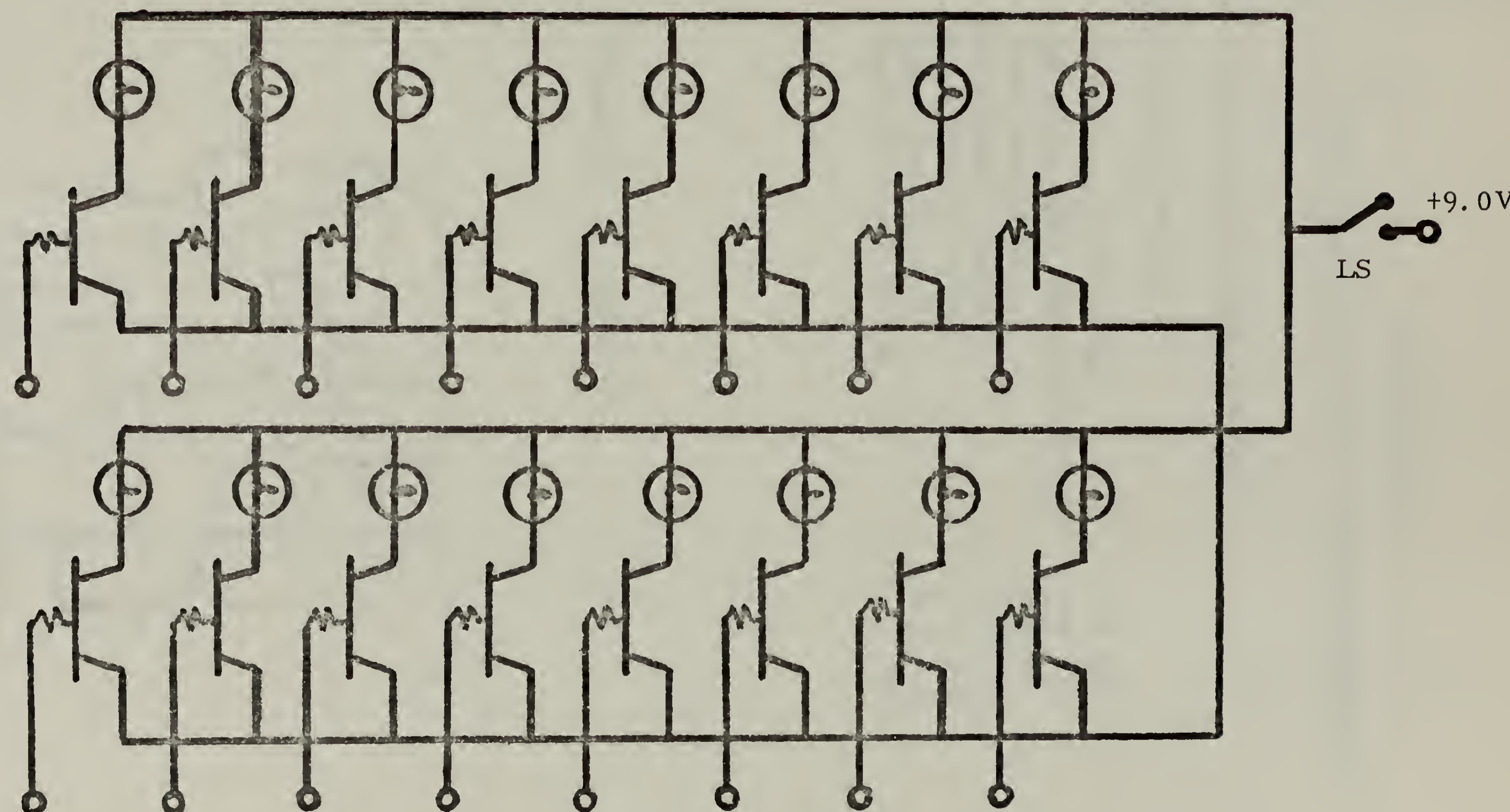
APPENDIX C OVERALL SCHEMATIC

COUNTER - 4 CuL9989



ALL RESISTORS = 10K
ALL TRANSISTORS = 2N3417
ALL LAMPS = MURALITE L10/20

OUTPUT UNIT



APPENDIX D

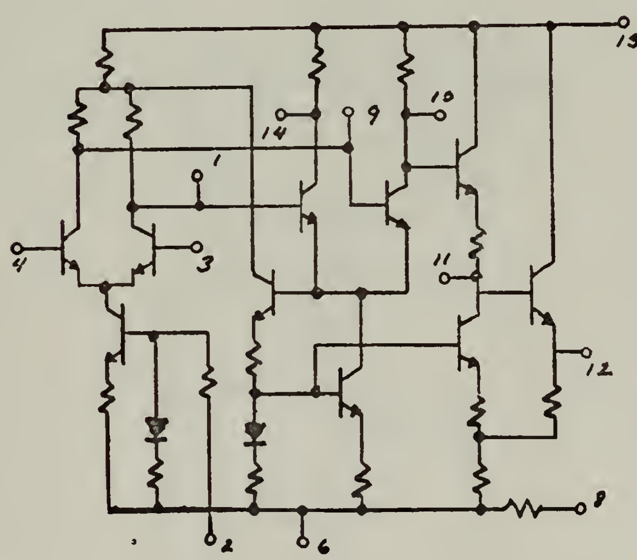
INTEGRATED CIRCUIT CHARACTERISTICS

A. RCA CA3029A Operational Amplifier

Typical Characteristics:

Open-Loop Voltage Gain.	60 dB
Common-Mode Rejection Ratio. .	94 dB
Input Impedance	20 kΩ
Input Offset Voltage.	0.9 mV
Input Offset Current	0.3 μA
Input Bias Current	2.5 μA
Static Power Drain at ± 3V . . .	7 mW

Schematic: CA3029A

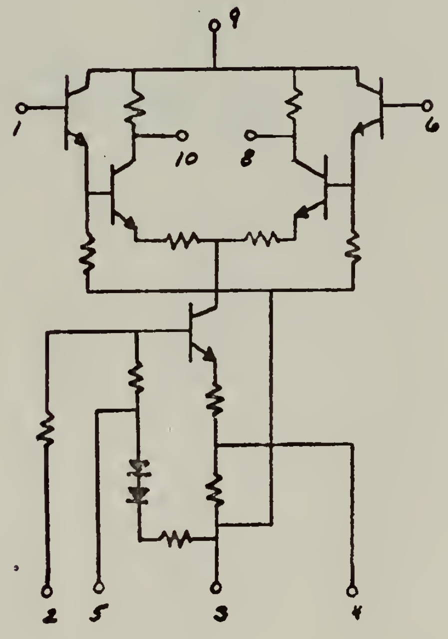


B. RCA CA3000 Differential Amplifier

Typical Characteristics:

Supply Voltages.	± 6 V
Device Dissipation	30 mW
Common-Mode Rejection	98 dB
Single Ended Input Impedance . .	195 k
Single Ended Ouput Impedance . .	5.5 to 10.5 k

Schematic: CA3000



C. Fairchild CuL9989 - 4-Bit Binary Counter

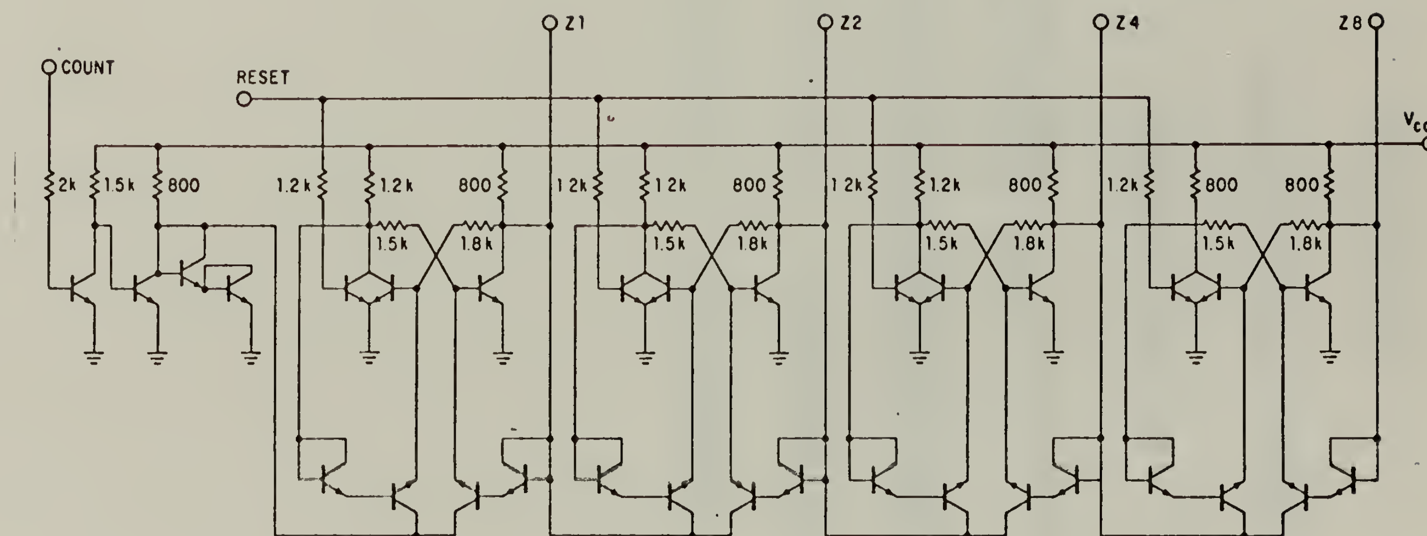
ABSOLUTE MAXIMUM RATINGS (above which life may be impaired)

Voltage at pin 7 ($0^{o}C$ to $+75^{o}C$) (TO-99)	+6.OV
Count Input Pin Voltage	+4.OV, -2.OV
Reset Input Pin Voltage	+4.OV, -2.OV
Current into Each Output Terminal	$\pm$5.0mA

ELECTRICAL CHARACTERISTICS (0-75^{o}C Free Air Temperature unless otherwise stated)

PARAMETER	MIN.	TYP.	MAX.	UNITS
Supply Voltage-V_{cc}	3.6		5.5	Volts
Power Dissipation		300	385	mW
Power Dissipation		132		mW
Count Input-Low-V_{ilc}			0.45	Volts
Count Input High-V_{ihc}	1.2			Volts
Count Input Current		210	330	uA
Count Input Pulse Width-High	40			ns
Count Input Slope-Positive Going		1.0		v/us
Max. Freq. of Input Count Pulses	10	15		MHz
Reset Input-Low-V_{ilr}			0.45	Volts
Reset Input-High-V_{ihr}	1.2			Volts
Reset Input Current		1.45	2.30	mA
Reset Input Pulse Width-High		220		ns
Output-Low-V_{ol}			0.45	Volts
Output-High-V_{oh}	1.2			Volts
Max. Delay From Count Input To Z_8 Output		90	120	ns

Schematic: CuL9989



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Unclassified
Security Classification

DOCUMENT CONTROL DATA - R & D

(Security classification of title, body of abstract and indexing annotation must be entered when the overall report is classified)

1 ORIGINATING ACTIVITY (Corporate author) Naval Postgraduate School Monterey, California 93940	2a. REPORT SECURITY CLASSIFICATION Unclassified 2b. GROUP

3 REPORT TITLE

A Heart Beat Accumulator For Research In Exercise Physiology

4 DESCRIPTIVE NOTES (Type of report and, inclusive dates)

Master's Thesis; June 1971

5 AUTHOR(S) (First name, middle initial, last name)

Richard A. Creighton

6 REPORT DATE	7a. TOTAL NO. OF PAGES	7b. NO. OF REFS
June 1971	44	6

8a. CONTRACT OR GRANT NO. b. PROJECT NO c. d.	9a. ORIGINATOR'S REPORT NUMBER(S) 9b. OTHER REPORT NO(S) (Any other numbers that may be assigned this report)

10 DISTRIBUTION STATEMENT

Approved for public release; distribution unlimited.

11. SUPPLEMENTARY NOTES	12. SPONSORING MILITARY ACTIVITY

13. ABSTRACT

A system is postulated to count the total number of heart beats in one day. The device is intended for use as a possible indicator of the level of physical fitness of an individual. It is portable, self-contained, and provides for comfortable and natural movement of the subject during the course of daily activities. Electronic design is developed including detailed schematic diagrams of all electronic circuitry involved. The complete plans and photographs of a working prototype are presented. The prototype is to be further tested and evaluated in the exercise physiology laboratory at California State College at Long Beach. It is hoped that this device will prove to be of value in studies and research in physical fitness.

KEY WORDS	LINK A		LINK B		LINK C	
	ROLE	WT	ROLE	WT	ROLE	WT
Physical Fitness						
Heart Beat Accumulator						

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